DEVELOPMENT OF KNEE JOINT ANGLE TRAJECTORY TRACKING FOR LEG SUPPORT MECHANISM

WAN MOHD SHUKRI BIN WAN SALLEH

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Universiti Tun Hussein Onn Malaysia

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ABSTRACT

Human assistive device is a device designed for helping human with disability and impairment problem to perform daily activities as normal person. The main categories of the assistive device are passive and active types. The active type requires power source to operate its mechanism while the passive type does not require power source. One of the important assistive devices is Knee-Ankle Foot Orthosis (KAFO) that designed for providing leg support during walking. This project is focused on the development of leg support mechanism consists of thigh and shank braces equipped with an actuator and gearing system purposely designed for providing additional support to the wearer during walking. This powered mechanism has ability of tracking the knee joint angle trajectory to imitate gait movement of a normal healthy person. The normal knee joint angle trajectory data is obtained from healthy person by using wireless Radio Frequency communication device and then plotted in Matlab Graphical User Interface (GUI) in a real-time measurement system. Proportional Integral Derivative (PID) control system is employed in this project as a controller to make sure the actuator of this mechanism moves according to the desired knee joint angle trajectory data. Designing a PID control system for knee joint angle trajectory tracking requires proper tuning of PID parameters to produce good output response. A step input is given to the system and its output response is observed while tuning these parameters. Then the actual desired knee joint angle trajectory data is given to the system and tuning process is repeated again to get best output response that shows closest angle trajectory compared to the desired trajectory.

ABSTRAK

Alat bantuan sokongan bagi manusia merupakan suatu alat yang direkabentuk bagi membantu orang kurang upaya dan pesakit yang mempunyai masalah kemerosotan anggota badan untuk melaksanakan aktiviti harian seperti orang yang normal. Kategori utama bagi alat bantuan sokongan ini ialah dari jenis aktif dan pasif. Kategori aktif memerlukan bekalan kuasa untuk menggerakkan mekanisme yang terlibat manakala kategori pasif tidak memerlukan bekalan kuasa. Salah satu alat bantuan sokongan yang penting ialah *Knee-Ankle Foot Orthosis (KAFO)* yang dicipta untuk menyediakan sokongan ketika berjalan. Projek ini difokuskan kepada pembangunan alat bantuan sokongan untuk lutut yang terdiri daripada pendakap aluminium di bahagian betis dan paha yang dilengkapi dengan penggerak bersama sistem gear yang bertujuan menyediakan sokongan tambahan kepada pengguna ketika berjalan. Mekanisme ini mempunyai kemampuan mengesan dan mengikuti trajektori sudut lutut untuk menghasilkan gaya berjalan yang sama dengan orang yang sihat. Data bagi sudut lutut yang normal ketika berjalan diperolehi daripada orang yang sihat menggunakan peranti komunikasi Frekuensi Radio tanpa wayar dan dipaparkan pada graf dalam perisian *Matlab Graphical User Interface (GUI)* secara pengukuran masa nyata. Sistem kawalan *Proportional Integral Derivative (PID)* digunakan dalam projek ini sebagai pengawal untuk memastikan penggerak bagi mekanisme ini bergerak berdasarkan data trajektori sudut lutut yang dikehendaki. Proses merekabentuk pengawal PID memerlukan penentuan nilai parameter-parameter PID untuk menghasilkan sambutan keluaran yang baik. Suatu isyarat langkah diberikan kepada sistem ini dan sambutan keluarannya diperhatikan sambil mengubah parameter-parameter tersebut. Kemudian data trajektori sudut lutut yang sebenar diberikan kepada system ini dan proses penentuan parameter diulang semula untuk mendapatkan sambutan keluaran terbaik yang menunjukkan trajektori sudut lutut yang paling hamper dengan trajektori yang dikehendaki.

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CHAPTER 1

INTRODUCTION

1.1 Project Background

The study and research about human assistive device or known as orthotics and prosthesis was started since late 1940s, as reported by [1]. The purpose of this device is to help human with disability and impairment problems to perform their Activities of Daily Living (ADL) as normal person such as eating, bathing, dressing and transferring. Examples of transfer activities are sit to stand and stand to sit (STS) movement, walking, stair ascent and stair descent. Many types of orthotics have been designed to assist patients that suffer from upper and lower limb disabilities due to many neuromuscular diseases such as Spinal Cord Injury (SCI), Muscular Dystrophy (MD) and Spinal Muscular Atrophy (SMA). The type of orthotics is categorized according to the supported parts of human body such as Ankle-Foot Orthotic (AFO), Knee-Ankle Foot Orthotic (KAFO), Hip-Knee Ankle Foot Orthotic (HKAFO) and Hip Guidance Orthotic (HGO). The operations of these orthotics are either passive, powered or by using functional electrical stimulation (FES).

This project is about the development of an assistive device to provide additional support for patients with lower limb disability and impairment problems to perform walking movement. The focus of this project is to introduce powered orthosis that able to provide additional force for the patient to perform leg flexion and extension during gait cycle according to the normal healthy person. The device is developed by using an actuator positioned at the knee joint and incorporating a PID control system to regulate the actuator speed for tracking the desired knee joint angle trajectory as a normal healthy person. An experimental method is conducted to obtain the characteristics of knee actuator and design the PID controller of the system. The performance of this assistive device is evaluated by using MATLAB simulation and real time testing on the hardware.

1.2 Problem Statement

Patients who suffer from lower limb impairment problems such as muscle weakness due to Spinal Cord Injury (SCI), muscular dystrophy and Post-polio syndrome have difficulty to perform leg flexion and extension during walking movement. An external physical support device is identified as one of beneficial method to assist the patients to perform both leg flexion and extension as a normal person, mostly by using active powered assistive device.

Active powered assistive device that were developed since years ago show a trend of incorporating a control system that able to control leg flexion and extension during walking movement. The purpose of the control system is to produce movement according to natural human behaviour, increase stability of the system, and reduce power consumption. Among the widely used control systems are PID, fuzzy logic, and genetic algorithm combined with sensory systems as feedback such as actuator encoder and Electromyogram (EMG) signal. However there is a big challenge of using this kind of human physiological signal as feedback which is requiring complex human body modeling algorithm. Therefore this project introduces a more simple yet convincing PID control system with external sensory system to infer intended walking motion of the user.

1.3 Aim and Objective of Project

The aim of this project is to develop human leg support mechanism to provide support during walking movement. In order to achieve the aim of this project, the objectives are outlined as follows:

i. To develop a prototype of leg support mechanism consists of thigh and shank braces equipped with knee actuator to provide leg support during walking movement.
ii. To obtain knee joint angle trajectory data for normal healthy person during walking movement.
iii. To develop PID control system for knee joint angle trajectory tracking and evaluate its performance.

1.4 Scope and Limitation

This project is intended to be used by patients with muscle weakness and incomplete paraplegic patients who still have minimum ability to perform walking movement with their own effort.

The project development is focused on the hardware and software implementation based on dynamics and control system theory of joint in the mechanism. The dynamics of the actuator and controller design are obtained from data-driven modeling approach of the system.

The performance of the device that to be evaluated is based on the specification and limitation of proposed mechanism by means of actuator types, control scheme, and sensory feedback method adopted in the system.

1.5 Organization of Thesis

This thesis is organized in five chapters that explain the theoretical aspect and development process of the project. These chapters are arranged in sequence order as follows:

Chapter II: Literature Review. This chapter discusses about studies and researches conducted by other scholars related to this project. The overview of history, comparison between various types of human support devices, and its summary of features are presented in this chapter.

Chapter III: Methodology. This chapter describes the approaches used throughout the development of this project which covers the hardware system of the device and software implementation to control the whole operations of the device.

Chapter IV: Result and Analysis. This chapter presents the findings, observation and data collections of this project. Each result is presented, analysed and discussed part by part based on the sub-topic.

Chapter V: Conclusion. This final chapter summarizes the result and analysis to obtain conclusion of this project with regards to the objectives that previously outlined. Future improvement and recommendation are presented also in this chapter as contribution for others to acquire benefits from this study.

CHAPTER 2

BRIEF REVIEW OF HUMAN SUPPORT DEVICES

2.1 Introduction

Many studies and researches have been conducted regarding the development of human support devices known as orthotics and prosthetics since late 1940s, as reported by [1]. An orthotic is defined as an externally applied device that is designed and fitted to the body in order to achieve control of biomechanical alignment, protect and support and injury, assist rehabilitation and increase mobility. While the term prosthesis refers to an artificial device attached to the body for replacement of missing part [2].

The early study of this area focused on the passive type of orthotics and prostheses, then upgraded to the active type by introducing powered mechanism of the devices. The early development of powered orthotic or known as robotic orthotic began since 1970s, introduced by Miomir Vukobratovic from the Mihailo Pupin Institute in Belgrade who constructed active lower limb assistive device by using hydraulic actuators to provide flexion and extension of the hip, knee and ankle [3]. Recently the term assistive orthotics also called as exoskeleton which means external skeleton that refers to the orthotics that fitted to external human body.

This chapter discusses about history, previous work and related researches done by other researchers regarding the assistive orthotic for lower limb. The research background, methods and results from various articles are analysed and commented accordingly in order to relate with this proposed project. This chapter presents also the rationale of conducting this proposed project by considering other researches that had been done previously.

2.2 Lower Limb Impairment Diseases

Many people around the world suffers from various diseases that result in lower limb impairment problems. The category of the diseases can be classified into three types which are peripheral neurological diseases (such as poliomyelitis and post-polio syndrome, spina bifida and poly nueropathy), muscular diseases (such as duchenne muscular dystrophy), and central neurological diseases (such as multiple sclerosis, cerebral palsy and spinal cord injury) [4]. According to [5], the diseases that may require use of assistive orthosis includes cerebral palsy, spina bifida, arthritis, diabetes, spinal cord injury, cerebral vascular accident, peripheral nerve injury, and various ligamenious and tendon injuries.

A brief explanation about selected diseases among the outlined previously are discussed in the following sub topic.

2.2.1 Poliomyelitis and Post-polio Syndrome (PPS)

Poliomyelitis is one of motor neuron diseases that related to neurological disorders. This acute viral disease is caused by a virus that attacks the human nervous system. People at any age are possible to be infected by this disease, however it mainly affects children under 5 years old [6].

Post-polio Syndrome (PPS) is a condition that occurs to the polio survivors years after recovery of initial acute attack of polio virus. The symptom of this syndrome includes gradual muscle weakness and decrease of muscle size or known as muscle atrophy. Consequently, this syndrome affects the patients' independency such as lost of mobility and ambulatory [7].

2.2.2 Muscular Dystrophy (MD)

Muscular Dystrophy (MD) is one of genetic diseases characterized by progressive weakness and degeneration of the skeletal muscles that control movement. In some cases, the symptoms of MD can be seen at early childhood while others are detected at middle age.

The effects of the muscle weakness due to MD include respiratory problem, functionality disorder, and loss of walking ability. The treatment for MD patients normally involves physical therapy, respiratory therapy, and use of physical orthopaedic appliance support as well as assistive devices.

2.3 Convention of Knee Joint Angle

The knee joint angle measurement used throughout this project is based on the convention shown in Figure 2.1.

Based on the figure, human lower extremities are comprises of segments which are hip, thigh, shank, ankle and foot. Normally there are two types of angle named as absolute angle and relative angle that used to describe the kinematics of human body segments. Knee joint angle is a type of relative angle measured between the longitudinal axis of two adjacent segments; knee and foot. Knee flexion refers to the movement of shank that increases knee joint angle while knee extension reduces knee joint angle.

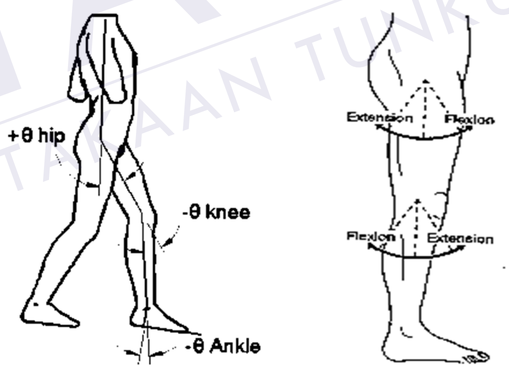

Figure 2.1: Lower Extremities Joints Angle

The knee joint angle can be measured either in degree or radian unit, however the unit chosen in this project is in degree unit. During standing posture, the thigh and shank are in straight extended position that results in the knee joint angle is 0 degree. This condition is considered as initial position of the knee joint.

2.4 Fundamental of Human Gait Analysis

Gait analysis is a study of biomechanics of human movement considering the parameters related to the functioning of lower extremities [8]. Referring to Figure 2.2, one cycle of gait is defined as the period between subsequent heel contact of the same foot. The cycle consists of stance phase and swing phase of a leg, where 60% of the gait cycle is made up of stance phase and the remaining 40% is swing phase. Stance phase means one of the foot is in contact with ground, while swing phase refers to the leg swings until next heel contact of the same foot.

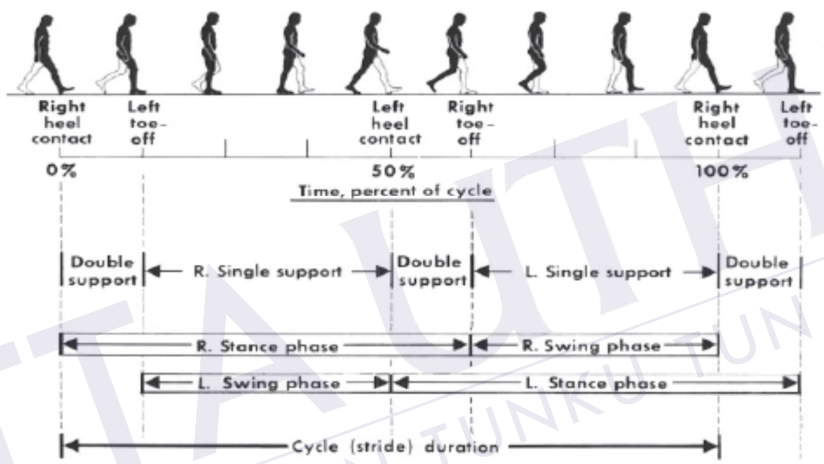

Figure 2.2: Phases in gait cycle [8]

The gait pattern of a human is unique and differs between a person and the other. There are many parameters characterizing the gait pattern of a person, among the parameters are stride time and length, step time and length, cadence, velocity and joints angle. The kinematic of gait in terms of displacement, velocity and acceleration are determined by factors including gender, walking speed, age, weight, height and body mass index.

A normal gait pattern shows knee angular displacement as shown in the following Figure 2.3, as described by [9]. The range of knee joint angle during walking movement is between 0° to 70° for a gait cycle. The knee flexion and extension are performed throughout the gait cycle starting with the first knee flexion happens between 10% to 20% of the gait cycle. At this moment, the knee flexion reaches 20° as the subject achieves foot flat condition. The knee begins to extend and reaches maximum extension at about 35% to 40% of the gait cycle as the heel rises from the ground. The knee flexes again for second time and reaches to maximum of 70° approximately at 65% of the gait cycle. Then knee extension continues until reaches to maximum just before ground contact.

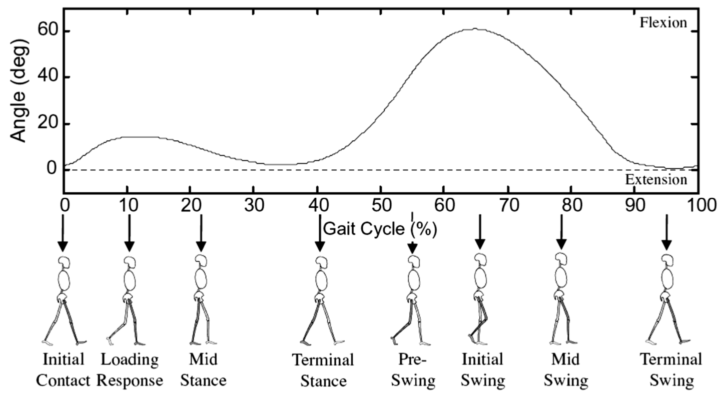

Figure 2.3: Knee joint angle of a gait cycle [9]

Normally a person with gait disorders exhibits deviations from the standard gait pattern such as maximum knee flexion and timing for each phase. The knee angular displacement pattern of a normal person is important as a guideline for generating trajectory data for the powered gait assistive orthotic. The trajectory data can be used as input to the orthotic system in order to drive the actuator according to the desired position throughout the gait cycle.

2.5 Overview of Assistive Orthotic and Exoskeleton

The research and development of assistive orthotic especially for lower extremities have been conducted years ago starting with passive orthotic that designed only for supporting the impaired limbs without providing additional force from active actuators. With the increase number of lower limbs impairment problems, there was a great demand for enhancement of previous passive orthotic to be powered with active actuator to extend the capability of the orthotic system. Recently, the active orthotic has emerged with the incorporating of a control system to provide a system that able to imitate natural human movement.

Examples of more advance human support devices or called as exoskeleton that have been developed are Hybrid Assistive Limb (HAL), Walkbot, RoboKnee, REX, ReWalk, Honda Walk Assist, eLEGS and Vanderbilt exoskeleton. HAL is invented by Yoshiyuki Sankai from Japan. This exoskeleton robot suit able to expand and improves physical capabilities of a human from healthy person to completely paraplegic patient. The purpose of HAL is mainly in industry for lifting weights and for medical purposes such as rehabilitation. It is considered as the most advance exoskeleton since it has specialized control algorithms named as 'Cybernic voluntary control' and 'Cybernic autonomous control'. Cybernic voluntary control is used to control the actuator torque to augment joint torque of the wearer according to the voluntary muscle activity, while Cybernic autonomous control has the capability to support a functional motion that is desired by the wearer which is inferred from a preliminary motion before the desired motion is implemented [10]. Figure 2.4 shows the system configuration of HAL-5.

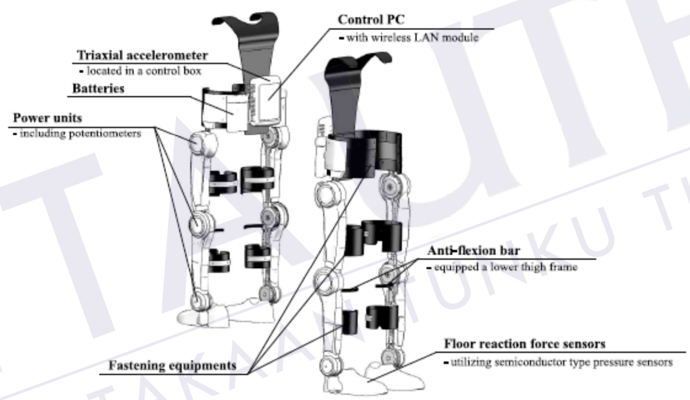

Figure 2.4: HAL-5 LB system configuration

Argo Medical Technologies, a company from Israel unveiled the assistive exoskeleton ReWalk in 2008. ReWalk was designed to provide walking movement for disabled person by actuating the knee and hip flexion/extension. The patients able to perform walking, stair ascent and descent, allows turning as well as sit and stand movement. However patients must use crutches to maintain their balance. Sensors located on the chest determine the angle of the torso and measure the patient's shift in gravity and upper body movements. The use of this exoskeleton is limited to patients who meet certain height and weight criteria [3].

Figure 2.5: ReWalk

Meanwhile, in 2010 Berkeley Bionics developed eLEGS, a lower limbs exoskeleton for the paralyzed patients to regain stand up and walk. eLEGS actively supports the knee and hip flexion/extension. The hip abduction/adduction is loaded with a stiff elastic component to minimize the unnatural posture, and it is equipped with spring-loaded ankle to reduce toe drop. It applied an artificial intelligence control combined with gesture recognition technologies realized by pressure sensors, potentiometers and accelerometer gyroscope board [3]. eLEGS users still need to use crutches to support the upper limb during walking. This exoskeleton suits for patients with heights between 157 to 193 cm and maximum weight of 100 kg, while the weight of this exoskeleton is 20 kg.

Figure 2.6: eLEGS

Vanderbilt powered orthotic was designed by Vanderbilt University (USA) to provide gait assistance for paraplegics by providing sagittal plane assistive torques at both hip and knee joints. The orthotic consists of thigh and shank braces equipped with DC brushless motor through a 24:1 gear reduction. The knee motors are equipped with electrically controlled normally locked brake. The knee joints are locked during stance phase of gait and released during sit to stand or stand to sit movements and during swing phase of gait [11].

Figure 2.7: Vanderbilt powered orthotic

The comparison between previous mentioned orthotics and exoskeletons in terms of type, actuator type, and method of control is summarized in the following Table 2.1.

Table 2.1: Comparison between gait assistive orthoses/exoskeleton

Previous research	Orthotic type	Actuator type	Method of Control
Vanderbilt (2011)	Hip Knee Ankle Foot Orthotic	DC brushless motor.	Use a finite state machine to control joints by defining trajectory for each state.
eLEGS (2010)	Hip Knee Ankle Foot Orthotic	DC motor	Artificial intelligence and gesture recognition to interpret user intention of movement.
ReWalk (2008)	Hip Knee Ankle Foot Orthotic	DC motor	Sensors on the chest determine angle of torso and detect shift of gravity and upper body movements.

Table 2.1 (continued)

Previous research	Orthotic type	Actuator type	Method of Control
HAL-5 (2005)	Full-body exoskeleton	Servo motors.	Intention-based walking support. The intention to perform walking is inferred by measurement of floor reaction force and shift of Center of Gravity (COG)

2.6 Leg Support Mechanism for Walking Assistance

Leg support mechanism for walking assistance is a type of partial lower limbs exoskeleton since it supports only the function of leg during walking movement. There were related works have been conducted that similar to the developed project in this thesis in terms of orthotic type and its purpose.

Christian Fleischer and Gunter Hommel (2006) developed a knee exoskeleton whose torque was controlled by Electromyography (EMG) Signal. The aim of this research was to provide additional torque at knee to perform flexion and extension movement by recognizing the intended motion of the user through EMG signal from the muscle [12]. The block diagram of this exoskeleton system is depicted in Figure 2.8

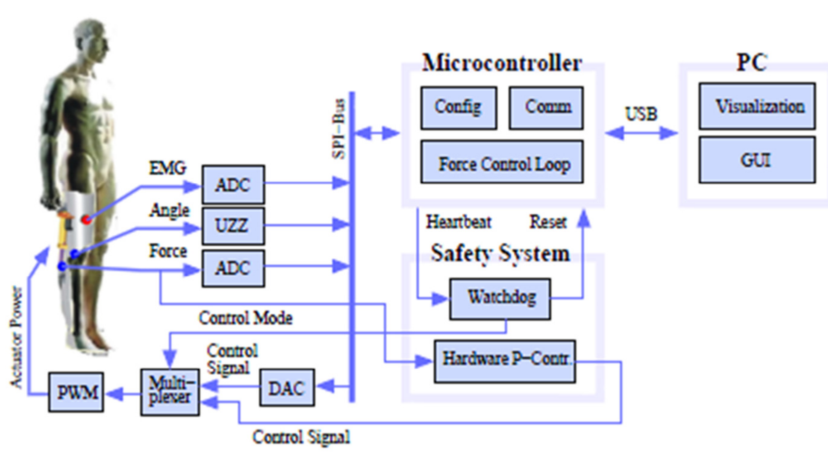

Figure 2.8: Exoskeleton knee torque control system

The exoskeleton consists of an orthotic attached to the right knee and powered by Maxon DC motor with ball screw shaft. The EMG signal activation is measured by using differential electrodes and sampled by 12-bit Digital to Analog Converter. The system uses Hall sensor to measure knee joint angle, while the force sensor is measured by force sensor. The microcontroller read EMG signal and knee joint angle to infer intended movement and then determine required knee torque for the movement. At the same moment, the actuator force and knee joint angle readings are used to calculate current knee torque produced by the actuator. The required knee torque is compared to the target (set point) of knee torque, then the controller decides either to increase, decrease or maintain the torque by giving PWM control signal to the actuator [12].

Another related work was done by Hongtao Guo *et al.* (2011) who conducted two studies regarding the knee brace orthotic as reported in [13] and [14] purposely to assist elderly or disabled people improving their mobility. This knee orthotic developed by Hongtao Guo *et al.* (2011) adopted a multifunctional actuator that comprises of two main parts which are the motor part and the clutch/brake part. A special fluids, magneto-rheological (MR) fluid is used inside the actuator to provide sufficient torque for the knee. The MR fluid and the clutch/brake part able to transfer the torque generated from the motor part to outside as a clutch or provide controllable semi-active torque as a brake with less power consumption than a conventional electric motor. The orthotic configuration is shown in Figure 2.9.

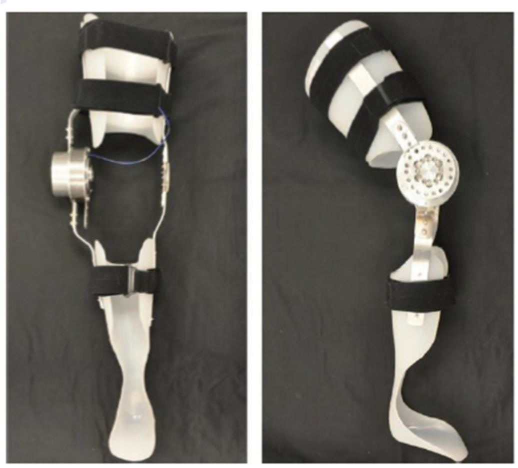

Figure 2.9: Knee-ankle-foot orthosis by Hongtao Guo *et al.*

In [13], a preliminary experimental study on kinematics and kinetics was carried out pertaining this orthotic to evaluate the changes between normal walking without using the orthotic and walking with the orthotic in 'off' mode. Hongtao Guo *et al.* found that the use of this orthotic in 'off' mode without actuation results in almost similar gait kinematics, kinetics, spatial and temporal parameters compared to the normal walking without the orthotic. Thus, they conclude that the design of the orthotic is suitable to be applied with assistive torque with a controlled actuator.

While in [14], Hongtao Guo *et al.* suggested an alteration of the previous orthotic to improve the knee flexion and rotation angle and also to find possible position of the actuator. The modification involved change of actuator position from subject's knee joint to hip joint, installation of hinge joint with bearing at the knee joint position, and one metal bar with two steel cables to form a linkage. This new version of orthotic is shown in Figure 2.10. The same experiments were conducted to evaluate the performance of this new version orthotic, however the results show that its performance is not improved compared to the earlier version.

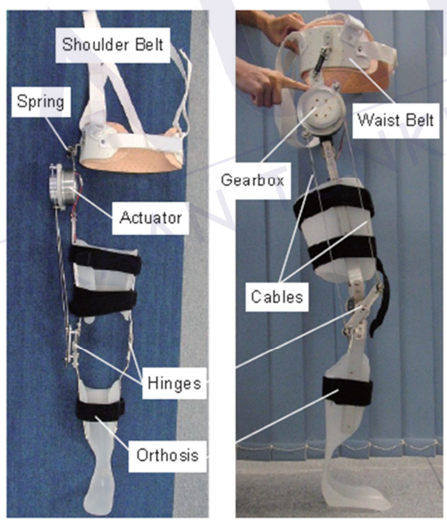

Figure 2.10: New version of Knee-ankle-foot orthosis by Hongtao Guo *et al.*

2.7 Design Considerations of Walking Assistive Orthotic

All of the powered orthotics described in this section have common design considerations and challenges prior to development of the devices to ensure its effectiveness in providing additional support for the user. The considerations include lightweight structure, safety, comfortable, portable and low power consumption, and high degree of reliability [15].

The mass of the orthotic has to be minimized to improve user comfort, reducing energy required to move the structure. A lower mass also results in low inertia during limb movements such as leg swing that eventually decrease the uncontrolled mechanical error.

As the device is applied directly to the human body, the safety is very crucial and topmost issue need to be considered. The device should not potentially harm the wearer or other neighbouring people either from electrical or mechanical aspects.

The user comfort means that the orthotic needs to have anthropomorphic design which suits to human body and able to imitate natural human movements. The orthotic also should not causing fatigue to the wearer even after long period of operation.

The portable and low power consumption of active actuators enables the user to use the orthotic for a long time period, however there is a trade-off between available power source such as battery and required actuator power required in the design.

An effective control system is important in order to provide high degree of reliability of the orthotic system. This includes accurate sensory feedback and fast output response of the control system. Normally this effective control system can be achieved by deploying advance algorithm such as artificial intelligent.

CHAPTER 3

METHODOLOGY

3.1 Introduction

This chapter describes the methodology used in the development of this project.

The method for developing this project is divided to two main sections. The first section is about the hardware identification for developing the overall system, and second section is about controller design approach involving use of MATLAB Simulink and microcontroller software.

3.2 Project Development

The development of this project is divided into several phases which covers hardware development, actuator characteristics identification and implementation of control system algorithm. These phases are shown by the following flowchart of Figure 3.1.

This project development begins with the development of leg support mechanism. Upon completion of the mechanism, it is used to obtain knee joint angle trajectory from healthy subjects. Then, the actuator characteristics in terms of relationship between speed and voltage are obtained through experimental method. The final phase is a real-time implementation of the system to the human and performance evaluation of the system.

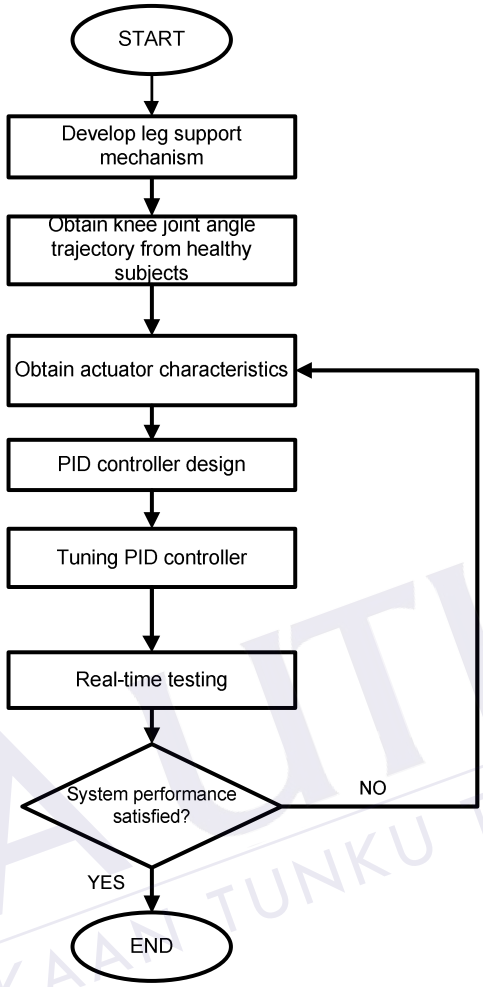

Figure 3.1: Project Development Flowchart

3.3 System Description

The system description for overall leg support mechanism is described by the following block diagram as shown in Figure 3.2 . The system consists of following parts; leg support mechanism, knee joint angle trajectory data as input, a Proportional Integral Derivative (PID) control system algorithm, potentiometer as feedback sensor, and Pulse Width Modulation (PWM) signal for controlling actuator speed. Details of each part are explained in next sub-topic.

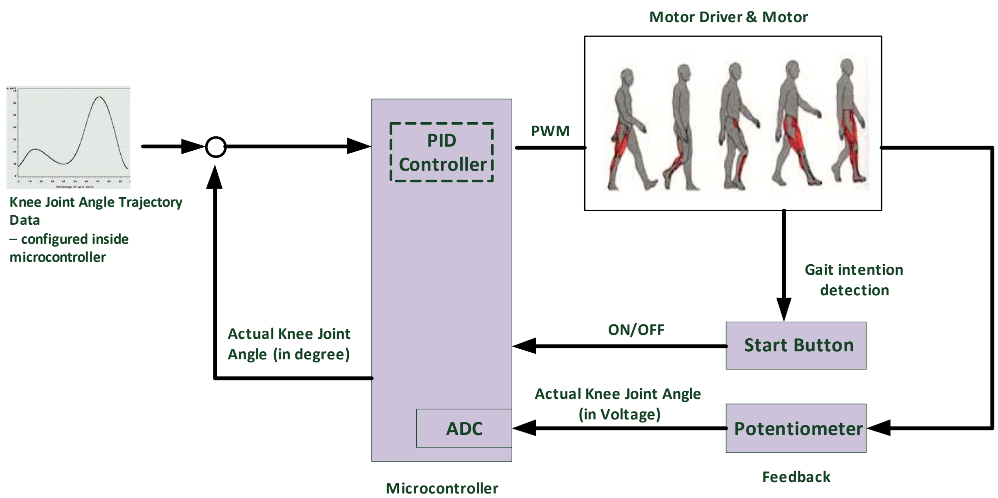

Figure 3.2: System Block Diagram

3.3.1 Leg Support Mechanism

The leg support mechanism of this project consists of two braces made of aluminium that cover thigh and shank at the peripheral side. Both braces are attached with a thermoplastic splint for supporting backside of thigh and shank. The overall knee orthotic system can be represented as 2 links robot with 1-DOF joint for one leg.

The knee joint actuator is positioned at lateral side of the knee. A pair of bevel gear is assembled to the motor shaft and shank brace to provide 90^{o} change of drive direction and increase the torque output at the knee joint. Figure 3.3 shows the structure of this mechanism.

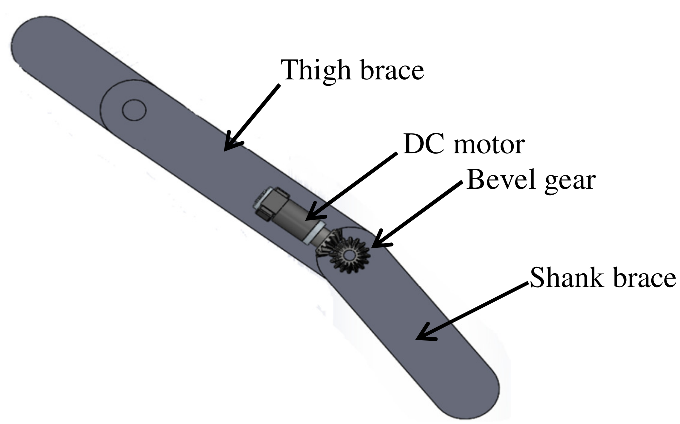

Figure 3.3: Leg Support Mechanism with Actuator and Bevel Gear

3.3.2 Knee Joint Angle Trajectory

The input of the system is knee joint angle trajectory data that gathered through experimental method on normal healthy subjects. The selected subjects with certain range of age and weight are selected to provide knee joint angle trajectory during gait cycle by wearing the leg support mechanism without activating the actuator.

The knee joint angle is measured by using a potentiometer attached to the orthotic at backside of the knee. An analog output voltage from the potentiometer is measured during knee flexion/extension that equivalent to the knee joint angle displacement. This analog voltage is then read by a microcontroller and converts it to digital form by using internal Analog to Digital Converter (ADC) module for further analysis.

In order to obtain relationship between output voltage of the voltage divider circuit and knee joint angle, a calibration need is conducted by using an angle measurement tool which is fixed to the braces. This measurement tool is a protractor scale that able to measure rotation angle up until 180^{o} with 1^{o} resolution.

3.3.3 DC Actuator

The actuator used in this project is DC planetary geared motor that operated with 12V DC source and rated 5.5 Amp current. This planetary geared motor has gearbox reduction ratio of 1:104, producing rated torque of 1.96N.m and rated speed of 63 rpm.

The motor speed is controlled by Pulse Width Modulation (PWM) signal from microcontroller through a motor driver circuit. The speed is proportional to the duty cycle of PWM signal whose value is determined by a control system developed in the microcontroller. Figure 3.4 and Figure 3.5 show the DC motor and it's characteristic.

Figure 3.4: DC planetary geared motor

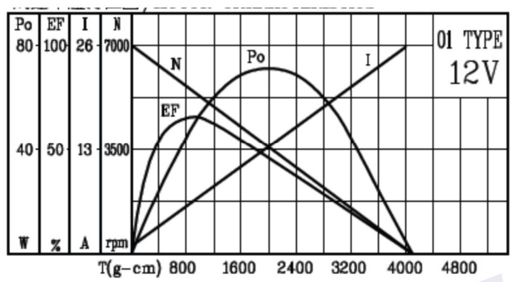

Figure 3.5: DC planetary geared motor characteristic

3.3.4 Motor Driver Circuit

A motor driver circuit is required to drive the DC motor from microcontroller. The selection of motor driver circuit is based on maximum current and motor control method. The selected motor driver circuit to be used in this project is MD10C model from Cytron that able to support motor voltage rating between 5V to 25V and continuous current of 30A peak.

This motor driver circuit provides speed control by using PWM signal with frequency up to 20 kHz. The PWM signal is generated by microcontroller and connected to this motor driver circuit through PWM input pin.

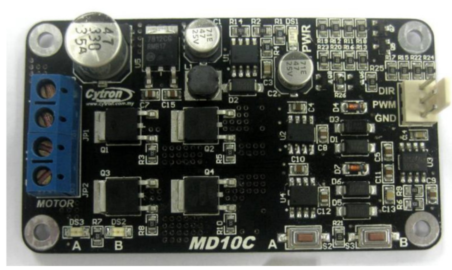

Figure 3.6: Motor driver circuit MD10C

3.3.5 Arduino Microcontroller Board

The Arduino Uno microcontroller board is used in this project. It is powered by ATmega328 microcontroller used for controlling the overall functions of the leg gait support mechanism system.

The main specifications of this Arduino Uno board that makes it suitable for this project are small size, operates with 5V DC source, has Analog to Digital internal (ADC) module, 6 channel of PWM signal and interrupt function. The availability of ADC, PWM and interrupt modules enable the implementation of PID control system algorithm by using C programming language embedded into the microcontroller.

3.4 PID Control Design

A closed loop control system will be used in this project rather than open loop system since its purpose is to produce output according to the desired input with minimum error and higher stability. A closed loop control system exhibits less sensitive to noise, disturbance, and changes in the environment [16].

Figure 3.7 describes the PID control block diagram for tracking knee joint angle trajectory. A PID control is proposed in this project which consists of Proportional, Integral and Derivative terms. Each of this term has constant gain which is K_P, K_I, and K_D. The output of PID controller, u(t) is the summing result of proportional, integral and derivative terms as shown in Equation 3.1.

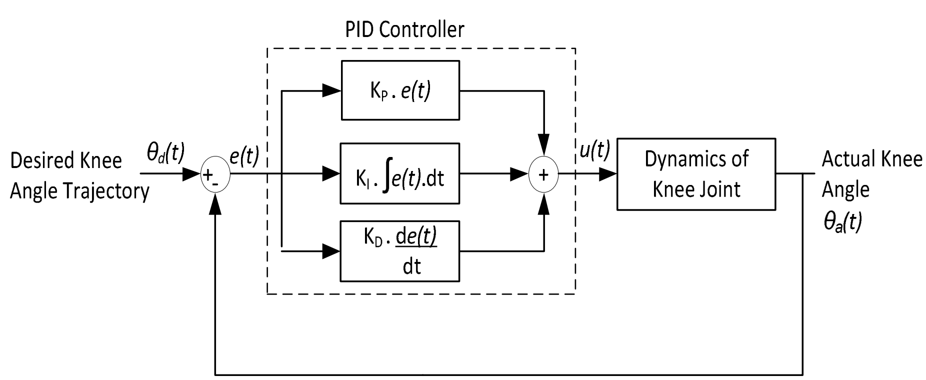

Figure 3.7: Block diagram of PID control system for knee joint angle tracking

$$u(t) = K_P.e(t) + K_I.\int e(t)dt + K_D.\frac{de(t)}{dt} \tag{3.1}$$

Where *e(t)* is the error or difference between desired and actual knee joint angle.

The output response of knee joint angle control system is influenced by the controller constants, K_P, K_I, and K_D. Proportional gain, K_P will reduce the rise time and steady-state error of the system. The integral gain, K_I will eliminate steady-state error but might produces worse transient response. The derivative gain, K_D increases the system stability, reducing overshoot percentage, and improves the transient response of the system. A tuning process shall be conducted to find the most suitable value for K_P, K_I, and K_D in order to obtain the optimum output response of the system.

The PID controller tuning process can be conducted once the characteristics of the actuator are obtained. An experimental method is conducted to obtain characteristic of this actuator in terms of relationship between PWM value from microcontroller and the actuator speed. A program is constructed in Arduino board microcontroller to give different PWM output values. This PWM value is fed into motor driver board starting at 0 degree knee joint angle position and it is stopped when reaches 90 degree position. During movement of the actuator, the angle and captured time are continuously gathered and plotted in Matlab GUI. The same procedure is repeated by using other PWM values, and finally an analysis regarding PWM effect to the angular displacement and acceleration is conducted. From this experiment, a range of suitable PWM value from minimum to maximum is decided to be used in the PID controller.

3.5 System Operation Flowchart

The system operation of this leg support mechanism is summarized in the following flowchart of Figure 3.8. Upon activated by power ON the device, the overall system is initialized by repositioning the knee joint at default condition as the user standing up with both foots are in contact with the floor.

The user intention to initiate walking is triggered by pressing start button on the console box. The potentiometer measures knee joint angle value and compares with the desired trajectory data continuously. Once controller detects deviation from desired trajectory data, the microcontroller will adjust the actuator speed by regulating the PWM signal to attain the trajectory. The operation continues for next gait cycle and will stop when both force sensors under the heels are in contact with the floor.

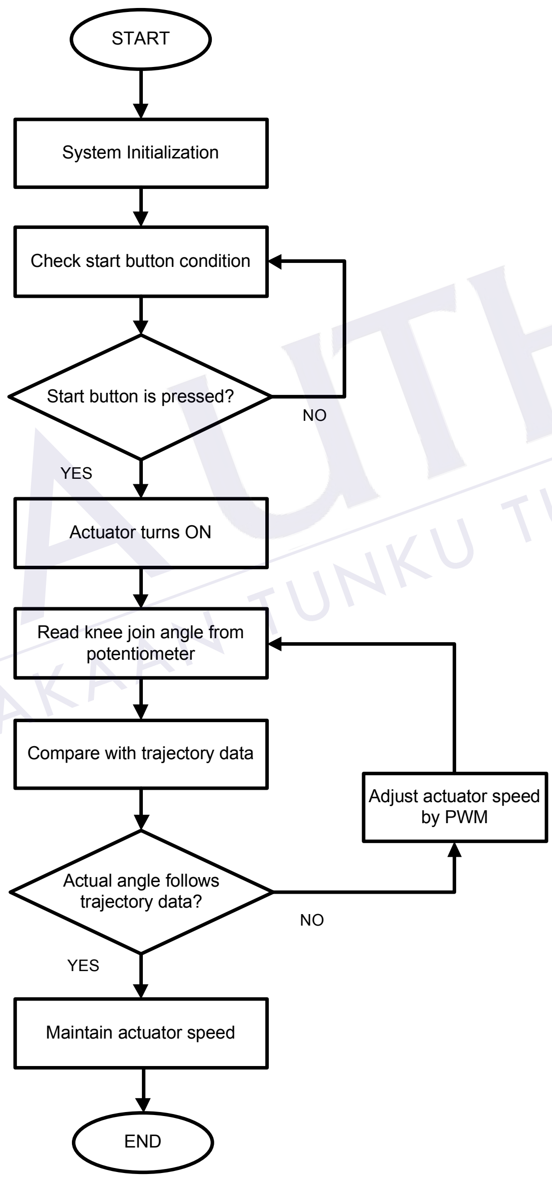

Figure 3.8: System Operation Flowchart

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